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## ORIGINAL ARTICLE

# Walking in an Unstable Environment: Strategies Used by Transtibial Amputees to Prevent Falling During Gait

Laura Hak, MSc,<sup>a</sup> Jaap H. van Dieën, PhD,<sup>a</sup> Peter van der Wurff, PhD,<sup>b,c</sup>  
Maarten R. Prins, BSc,<sup>a,b</sup> Agali Mert, PhD,<sup>b</sup> Peter J. Beek, PhD,<sup>a</sup> Han Houdijk, PhD<sup>a,d</sup>

From the <sup>a</sup>Research Institute MOVE, Faculty of Human Movement Sciences, VU University Amsterdam, Amsterdam; <sup>b</sup>Center for Augmented Motor Learning and Training, National Military Rehabilitation Centre Aardenburg, Doorn; <sup>c</sup>Pain Centre Department of Anaesthesiology, University Medical Centre Groningen, University of Groningen, Groningen; and <sup>d</sup>Heliomare Rehabilitation Centre, Wijk aan Zee, The Netherlands.

## Abstract

**Objective:** To investigate which strategies transtibial amputees use to cope with challenges of gait stability and gait adaptability, and how these strategies differ from strategies used by able-bodied controls.

**Design:** Cross-sectional study.

**Setting:** An instrumented treadmill mounted onto a 6 degrees-of-freedom motion platform in combination with a virtual environment.

**Participants:** Transtibial amputees (n=10) and able-bodied controls (n=9).

**Interventions:** Mediolateral (ML) translations of the walking surface were imposed to manipulate gait stability. To provoke an adaptive gait pattern, a gait adaptability task was used in which subjects had to hit virtual targets with markers guided by their knees.

**Main Outcome Measures:** Walking speed, step length, step frequency, step width, and selected measures of gait stability (short-term Lyapunov exponents and backward and ML margins of stability [MoS]).

**Results:** Amputees walked slower than able-bodied people, with a lower step frequency and wider steps. This resulted in a larger ML MoS but a smaller backward MoS for amputees. In response to the balance perturbation, both groups decreased step length and increased step frequency and step width. Walking speed did not change significantly in response to the perturbation. These adaptations induced an increase in ML and backward MoS. To perform the gait adaptability task, both groups decreased step length and increased step width, but did not change step frequency and walking speed. ML and backward MoS were maintained in both groups.

**Conclusions:** Transtibial amputees have the capacity to use the same strategies to deal with challenges of gait stability and adaptability, to the same extent as able-bodied people.

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Falling and fear of falling are significant health problems for people who walk with a prosthesis.<sup>1,2</sup> Explanations for why amputees fall more often than able-bodied people are that amputees are less stable during steady-state walking,<sup>3,4</sup> are less responsive to mechanical balance perturbations, and have less ability to adapt their gait pattern to environmental changes.<sup>5,6</sup> To improve gait stability and adaptability during the rehabilitation of amputees, insight into the strategies used by amputees to optimize

these aspects of walking, as well as how these strategies differ from strategies selected by able-bodied people, is required.

In a previous study,<sup>7</sup> we found that able-bodied people who were confronted with continuous balance perturbations during walking decreased their step length, increased their step width and step frequency, and kept their walking speed constant in order to maintain gait stability. These adaptations caused an increase in the mediolateral (ML) and backward margins of stability (MoS), which seems to be a compensation for the decrease in local dynamic stability (LDS) caused by the perturbations. When able-bodied subjects additionally had to adapt their gait pattern to suddenly appearing cues, these people were able to maintain sufficient MoS despite the disturbing effect of both manipulations. They did so by

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decreasing step length and increasing step width, but without an increase in step frequency, and therefore with a decrease in walking speed. The absence of an increase in step frequency in the latter task could be explained by the fact that this would decrease the available time to respond to the presented cues.<sup>8</sup>

The overall gait pattern of people with a lower limb amputation differs from that used by able-bodied people. Amputees often walk with wider steps than able-bodied controls, which can be explained as a strategy to decrease the risk of falling, because wider steps cause an increase of the MoS in the ML direction.<sup>9,10</sup> The increase in ML MoS possibly compensates for the lower LDS found for amputees, compared with able-bodied people.<sup>3</sup> Moreover, a lower preferred walking speed, with a lower cadence and smaller step length, was found for amputees.<sup>9,11</sup> However, whether these adaptations really serve the purpose of limiting fall risk or other possible purposes such as minimizing the energy cost to compensate for the mechanical constraints for propulsion<sup>12,13</sup> has not been elucidated. Moreover, it is unclear whether people with a lower limb amputation really do or can select strategies similar to those used by able-bodied people when gait stability or adaptability is challenged.

The purpose of this study was to investigate which strategies transtibial amputees use to cope with challenges of gait stability and adaptability. This was done by using ML balance perturbations and a gait adaptability task, in which subjects had to hit virtual targets that were projected on a 2-dimensional screen, using virtual markers controlled by knee motion. The purpose of this task was to simulate a real-life situation that requires accurate and fast adaptations of the normal, stable gait pattern, with a limited response time—for example, to avoid an obstacle that suddenly appears. Subjects walked on a self-paced treadmill, which made it possible for them to continuously adapt their walking speed. The effects of the manipulations on walking speed, step frequency, step length, and step width were measured, to investigate which strategies amputees used and how these strategies differed from strategies used by able-bodied people.

## Methods

### Participants

Ten adult subjects with a unilateral transtibial prosthesis (mean age  $\pm$  SD, 38.8 $\pm$ 14.6y; mean height  $\pm$  SD, 1.83 $\pm$ .11m; mean mass  $\pm$  SD, 87.1 $\pm$ 10.3kg; 9 men) and 9 age-matched control subjects (mean age  $\pm$  SD, 37 $\pm$ 11.4y; mean height  $\pm$  SD,

1.73 $\pm$ .08m; mean mass  $\pm$  SD, 70.2 $\pm$ 10kg; 4 men) participated in this study. Amputees and able-bodied controls were respectively recruited from the patient population and the employees of the Military Rehabilitation Centre Aardenburg, Doorn, The Netherlands. Further characteristics of the amputees are reported in table 1. All amputees used their own prosthesis (6 amputees walked with a fixed foot and 4 amputees with a flexible foot) and were able to walk in daily life without any walking device for at least 30 minutes. A minimum score of E on the Special Interest Group in Amputee Medicine mobility scale was necessary to participate in this study.<sup>14</sup> This study was approved by the medical ethical committee, and all subjects gave their written informed consent in accordance with university policy.

### Equipment

All subjects walked in the Computer Assisted Rehabilitation Environment (CAREN).<sup>a</sup> The CAREN system consists of an instrumented treadmill mounted onto a 6°-of-freedom motion platform in combination with a virtual environment (VE) (fig 1A). Twelve high-resolution infrared cameras<sup>b</sup> were used to capture kinematic data of 16 reflective markers attached to the pelvis and lower extremities (lower body plug-in-gait).<sup>15,16</sup> The treadmill was used in the self-paced mode, which allowed subjects to modify walking speed at will. This was done by servocontrolling the motor with a real-time algorithm that took into account the pelvis position in the anteroposterior (AP) direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling but did not provide weight support.

### Protocol

#### Warming up

Before the protocol started, subjects performed 5 warming-up trials of 3 minutes each, to become familiar with walking on a (self-paced) treadmill, the VE, and the various manipulations.

#### Experimental trials

The actual protocol consisted of 3 trials of walking 4 minutes at a self-paced walking speed: (1) a trial of unperturbed walking; (2) a trial with continuous balance perturbations; and (3) a trial with a gait adaptability task. These trials were offered in a random order.

For the balance perturbations, translations of the walking surface in the ML direction were used, following a multisine function (fig 1B):

$$D(t) = .05[1.0\sin(.16 * 2\pi t) + 0.8\sin(.21 * 2\pi t) + 1.4\sin(.24 * 2\pi t) + 0.5\sin(.49 * 2\pi t)] \quad (1)$$

where  $D(t)$  is the translation distance (m) and  $t$  is time (s).<sup>7,17,18</sup>

For the gait adaptability task, the VE was used to project targets on the screen (fig 1C). In addition, a projection of the markers attached to the knees was shown on the screen. Subjects were instructed to hit the targets on the screen with the projected knee markers, as close as possible to the center of the targets. A reason to choose for this task instead of a virtual obstacle avoidance tasks is the impossibility of stepping over virtual objects that are projected on a 2-dimensional screen. Stepping over virtual

#### List of abbreviations:

AP	anteroposterior
BoS	base of support
BW	backward
CAREN	Computer Assisted Rehabilitation Environment
CoM	center of mass
LASI	left anterior superior iliac spine
LDS	local dynamic stability
LPSI	left posterior superior iliac spine
$\lambda_{s-step}$	short-term Lyapunov exponent
ML	mediolateral
MoS	margins of stability
RASI	right anterior superior iliac spine
RPSI	right posterior superior iliac spine
VE	virtual environment
XCoM	extrapolated center of mass

**Table 1** Subject characteristics for amputees

Subject No.	Cause of Amputation	Side of Amputation	Time Since Amputation (mo)
1	Complex regional pain syndrome	Left	144
2	Traffic	Right	48
3	Traffic	Left	96
4	Trauma	Right	24
5	Blast	Left	27
6	Blast	Right	17
7	Vascular	Left	12
8	Traffic	Right	9
9	Traffic	Left	120
10	Blast	Right	34

objects would require a 3-dimensional environment. Another advantage of the adaptability task used in the current study is the possibility to quantify the performance on the task in terms of accuracy of the knee movement, while for an obstacle avoidance task only a pass and a hit can be distinguished from each other to quantify the performance. In each trial, a total of 32 targets appeared. Targets appeared at initial contact and disappeared after the duration of 1 gait cycle. The positions of the targets differed randomly in side (left or right), height (120% or 140% of lower leg length), and ML position (120% or 140% of distance between the left and right anterior superior iliac spines [LASI and RASI] from the midline of the treadmill).<sup>8</sup>

## Data collection

To calculate walking speed, the speed of the treadmill was recorded. Kinematic data of markers attached to the lateral malleoli of the ankles, the pelvis (LASI and RASI), and left and right posterior superior iliac spines (LPSI and RPSI), and the lateral epicondyles of the knees were collected with the Vicon system. The sample rate of data collection was 120 samples per second. The final 3 minutes of each trial was used for data analysis. Before data analysis, both speed data and kinematic data were low-pass filtered with a fourth-order bidirectional Butterworth filter with a cutoff frequency of 10Hz. However, this was not done for the calculation of the LDS, given the difficulties associated with filtering in the calculation of LDS.<sup>19</sup>

## Data analysis

### Walking speed

Walking speed was calculated as the average treadmill speed over the final 3 minutes of each trial.

### Step parameters

Step frequency was determined as the inverse of the average duration between 2 subsequent heel strikes, where heel strikes were detected as the local maxima of the position of the ankle markers in the AP direction. Step width was calculated as the ML distance between both ankle markers at the instant of heel contact, and step length was defined as the AP distance between these markers at the instant of heel contact.

## Gait stability

To quantify gait stability, LDS and the MoS in ML and backward (BW) direction were calculated.

LDS, a concept derived from nonlinear dynamics, can be quantified by the short-term (over 0–1 step) Lyapunov exponent ( $\lambda_{s-step}$ ), and is a measure for the attenuation of small perturbations that naturally occur during walking.<sup>20</sup> The  $\lambda_{s-step}$  expresses the logarithmic rate of divergence after a small disturbance of nearby orbits in a state space constructed from the markers placed on LASI, RASI, LPSI, and RPSI.<sup>7</sup> Negative exponents indicate local stability and positive exponents indicate local instability, with larger exponents indicating greater sensitivity to local perturbations.

The MoS were calculated, following a method derived from the method introduced by Hof et al,<sup>10,21</sup> as the difference in ML and BW direction between the extrapolated center of mass (XCoM) and the margin of the base of support (BoS) (fig 2). The XCoM is a concept that takes both the position and the velocity of the center mass (CoM) into account. Although basically similar, our method differs from that of Hof,<sup>10,21</sup> who used forceplate data for calculating the trajectory of the CoM and the XCoM. In the current study, the average of the pelvis markers was used to estimate the position of the CoM. The markers attached to the ankles were used to define the margin of the BoS.

## Gait adaptability

Gait adaptability was quantified by the performance on the gait adaptability task. This performance is defined as the minimum distance between the knee and the target center. For the period in which the target was visible on the screen, the minimal Euclidean distances between the knee markers and the center of the target in the plane of projection of the VE was assessed for each projected target (see fig 1C). The average of these distances was taken to get an outcome measure for the accuracy of the knee movements in performing the gait adaptability task.

## Statistical design

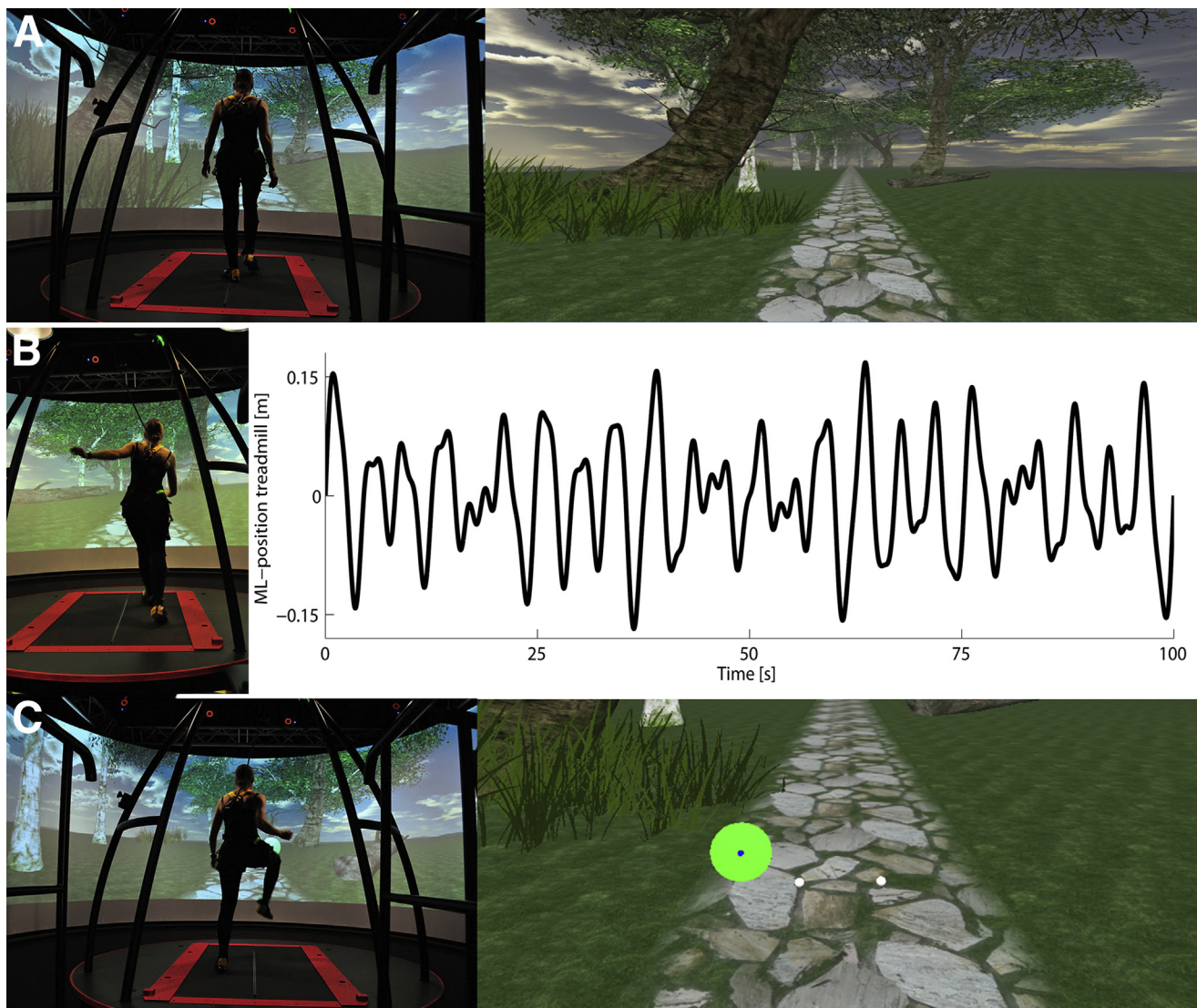
To measure the effects of the balance perturbation or the gait adaptability task on step length, step frequency, step width, walking speed,  $\lambda_{s-step}$ , and MoS, and to investigate whether these effects differ between amputees and healthy controls, 2×3 factorial analyses of variance were performed. The 3 conditions (normal walking, perturbed walking, walking with gait adaptability task) were used as within factor and group as between factor. Simple contrasts were used to determine for which condition (perturbed walking or walking with gait adaptability task) the variable concerned differed from normal walking. A paired samples *t* test was performed to investigate whether the performance on the gait adaptability task differed between both groups. Statistical analyses were performed using SPSS 17.0.<sup>c</sup>

## Results

One of the participating amputees (no. 7) was not able to complete the trials with the perturbation and the gait adaptability task. Therefore the analyses were done for 9 amputees and 9 healthy controls. The data of these remaining subjects did not contain any missing values.

Figure 3 shows the averages and SDs for walking speed, step frequency, step length, and step width for both groups and all 3





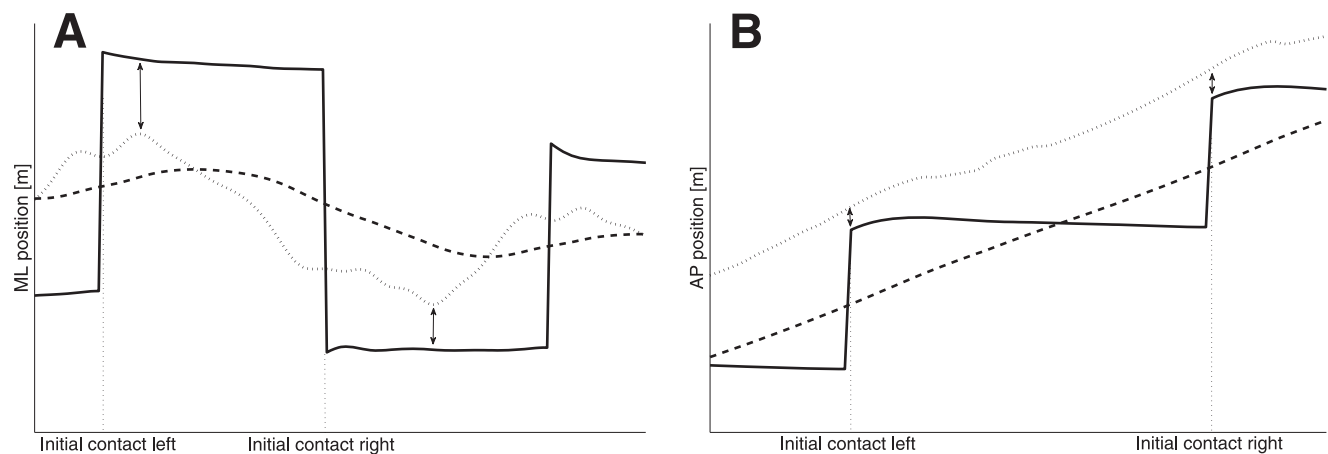
**Fig 1** (A) Experimental setup: CAREN and virtual scene. (B) ML balance perturbation with the perturbation pattern in the right panel. (C) Gait adaptability task with an example of a target in the right panel. The white dots represent a projection of the knee markers. The distance between the knee marker, in this example the left knee marker, and the center of the target was used as an outcome measure for the accuracy of the knee movements while performing the gait adaptability task.

conditions. Amputees walked on average slower than healthy controls ( $F_1 = 7.468$ ;  $P = .015$ ). Step frequency was significantly lower in the amputee group compared with the healthy group ( $F_1 = 11.427$ ;  $P < .01$ ), but step length did not differ significantly between groups ( $F_1 = 1.564$ ;  $P = .229$ ). Step width was larger for the amputees than for the healthy controls ( $F_1 = 7.503$ ;  $P = .015$ ). Values for  $\lambda_{s\text{-step}}$  were significantly higher for amputees than for healthy controls, which means that the amputees were locally less stable than the able-bodied controls ( $F_1 = 5.476$ ;  $P = .033$ ) (fig 4A). BW MoS were smaller for amputees than for able-bodied people ( $F_1 = 6.728$ ;  $P = .020$ ) (fig 4B). In contrast, amputees walked with larger ML MoS ( $F_1 = 7.774$ ;  $P = .013$ ) (fig 4C).

In response to the mechanical balance perturbations, both groups increased step frequency ( $F_1 = 26.078$ ;  $P < .01$ ) and step width ( $F = 37.028$ ;  $P < .01$ ), and decreased step length ( $F_1 = 23.223$ ;  $P < .01$ ), while walking speed did not change

significantly ( $F_1 = .046$ ;  $P = .832$ ). For both groups,  $\lambda_{s\text{-step}}$  increased ( $F_1 = 16.886$ ;  $P < .01$ ). Moreover, ML ( $F_1 = 22.190$ ;  $P < .01$ ) and BW MoS ( $F_1 = 21.151$ ;  $P < .01$ ) increased for both groups in response to the perturbations. No significant group by perturbation interactions were found.

During the trials with the gait adaptability task, both groups decreased step length ( $F_1 = 7.177$ ;  $P = .016$ ) and increased step width ( $F_1 = 85.967$ ;  $P < .01$ ), but step frequency ( $F_1 = 2.004$ ;  $P = .176$ ) and walking speed did not change significantly ( $F_1 = 2.203$ ;  $P = .157$ ). Both ML ( $F_1 = .023$ ;  $P = .881$ ) and BW MoS ( $F_1 = 2.425$ ;  $P = .139$ ) were not affected significantly by the gait adaptability task. Also, for the gait adaptability task, none of the measured variables showed a significant task by group interaction. The average distance  $\pm$  SD between the knee and the target, as the outcome measure of performance on the gait adaptability task, was  $3.71 \pm 1.30$  cm for the amputees and  $3.78 \pm 1.06$  cm for the able-bodied people, which was not significantly different from each other ( $t_{16} = .220$ ;  $P = .829$ ).



**Fig 2** Schematic representation of the calculation of the ML (A) and AP (B) MoS. Trajectories of the margin of the BoS (solid line), CoM (dashed line), and XCoM (dotted line) are shown for a period of approximately 2 steps. The XCoM is calculated as the position of the CoM plus its velocity times a factor  $\sqrt{l/g}$ , with  $l$  being the maximal height of the origin of the pelvis and  $g$  the acceleration of gravity. The MoS is the difference between the trajectory of the XCoM and the margin of the BoS for the moment at which the MoS reached its minimum value within the period of 1 step (represented by the arrows).

## Discussion

The purpose of this study was to investigate which strategies trans-tibial amputees use to cope with challenges of gait stability and gait adaptability, and how these strategies differ from those used by able-bodied controls. Despite differences in gait pattern during unperturbed walking, both groups responded in a similar fashion to the manipulations of stability and adaptability. Both groups increased step frequency and step width, decreased step length, and kept walking speed constant in response to the balance perturbation. To perform the gait adaptability task, both groups decreased step length, but did not change step frequency and walking speed significantly. In addition, an increase in step width was found in both groups. These results are in agreement with the results of previous studies<sup>7,8,17</sup> in which gait stability and adaptability were manipulated in able-bodied people.

In response to the balance perturbations, LDS decreased for both amputees and able-bodied people, as reflected by the increase of  $\lambda_{s-step}$ , but this decrease was compensated by an increase of the ML MoS and the BW MoS in the backward direction, which is a direct consequence of the adaptations in the different spatiotemporal gait parameters. An increase in BW MoS directly implies a decrease in the forward MoS. However, as was found previously, decreasing the risk of a backward fall appears to be preferred over decreasing the risk of a forward fall when stability of walking is challenged.<sup>7,22</sup> To prevent a backward fall, the XCoM should always be in front of the dorsal border of the BoS.<sup>23,24</sup> As the model of Pai and Patton<sup>24</sup> shows, the decrease in step length, in combination with the unchanged walking speed found in the present study, caused the increase in the BW MoS in response to the perturbations. In the ML direction, subjects should prevent that the XCoM exceeds the lateral border of the BoS. Hof et al<sup>10</sup> demonstrated that increasing step width and step frequency contributes to an increase in ML MoS. These adaptations were found in both groups in this study.

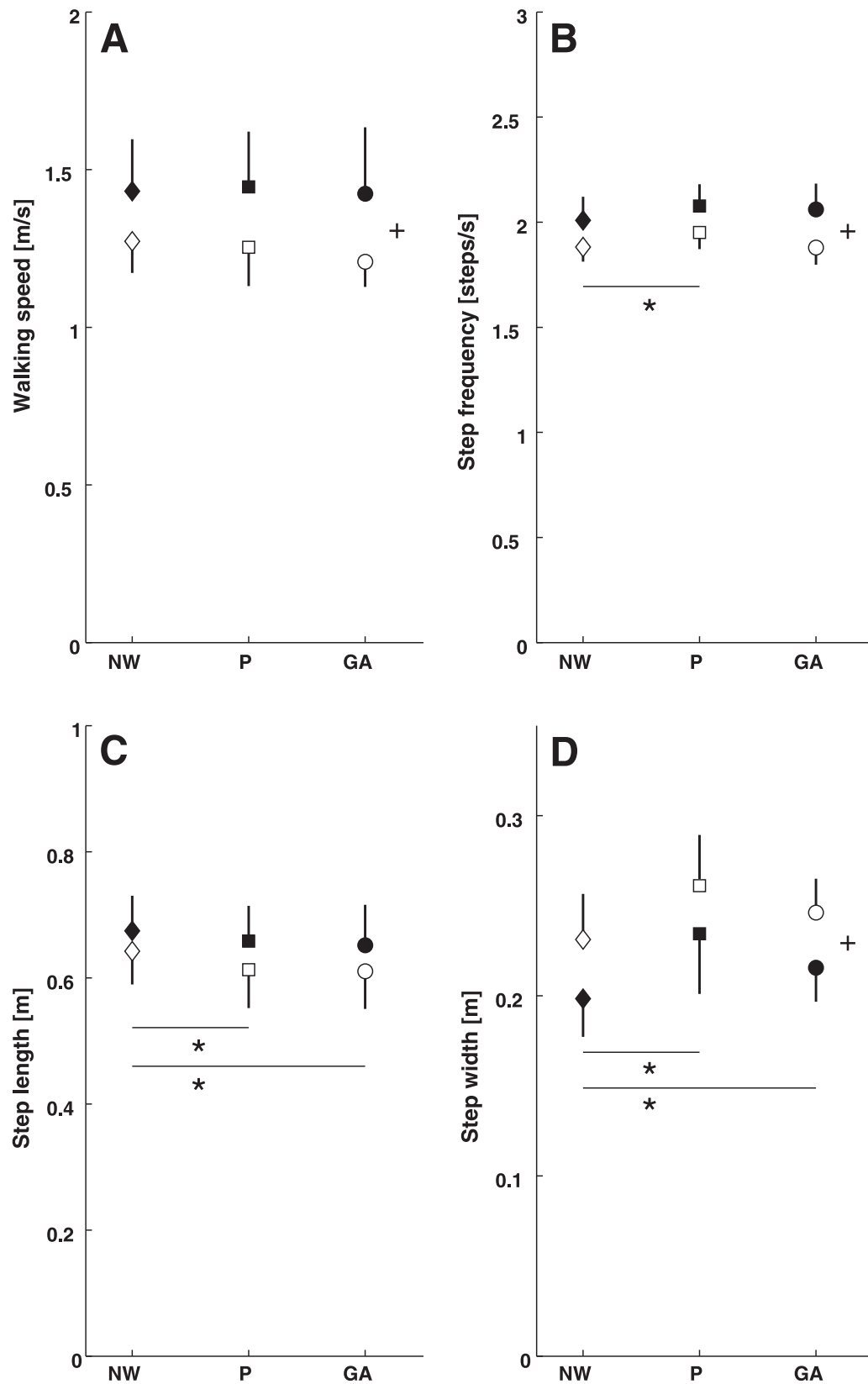
In situations that require fast adaptations of the gait pattern in response to environmental cues, it is important that these adaptations are adequate without losing balance.<sup>25-27</sup> The results showed that both amputees and able-bodied controls were capable of maintaining the ML and BW MoS during the gait adaptability task,

despite the disturbing effect of this task. To maintain the MoS, subjects increased step width and decreased step length. In contrast with the response to the balance perturbation, subjects did not increase step frequency to perform the gait adaptability task. In all likelihood, this was because an increase in step frequency would decrease the available time to respond to the targets, which would have had a negative influence on the accuracy of the hitting movement of the knee.<sup>28</sup>

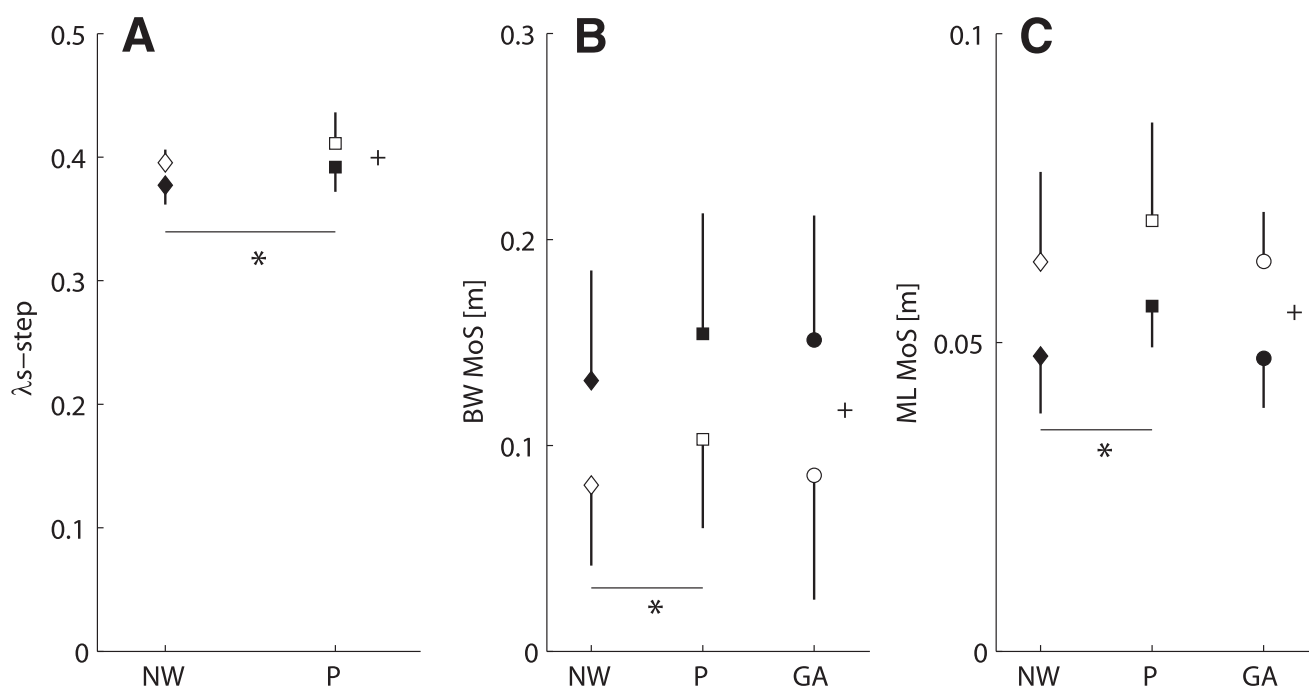
Although the response to the manipulations of gait stability and adaptability did not differ between amputees and healthy controls, indicating similar capacity and strategy to cope with perturbations and environmental cues, there were some overall differences between both groups. It is of interest to know whether these differences might be explained as a strategy to enhance stability and adaptability in people with an amputation, or serve other purposes. The LDS for amputees was lower compared with the able-bodied people, as reflected in the higher  $\lambda_{s-step}$  values for the amputee group. The larger step width for amputees, which resulted in a larger ML MoS, might be a strategy to compensate for this. On the other hand, the BW MoS were overall smaller for amputees than for able-bodied people, which could be taken to imply that amputees were unable to increase the BW MoS and consequently ran a larger risk of falling backward. However, this is unlikely because the amputees, just as able-bodied controls, increased their BW MoS when balance was perturbed. The smaller BW MoS could therefore be considered as a detrimental effect of the overall lower walking speed, caused by the lower step frequency used by amputees.<sup>23,24</sup> This lower step frequency could be an adequate adaptation to the adaptability manipulation, but this does not explain why the amputees also walked with a lower step frequency during the normal walking condition. Plausible reasons for the overall lower step frequency, and thus the lower walking speed, are the limited pushoff power and the higher energy demands of walking with a prosthesis compared with normal walking.<sup>12,13</sup>

## Study limitations

An important limitation is the small number of subjects that were included in this study, which might have affected the power of the



**Fig 3** Average and SD of walking speed (A), step frequency (B), step length (C), and step width (D) for amputees (white symbols; n=9) and healthy controls (black symbols; n=9). Abbreviations: NW, normal walking; P, perturbation. +Significant group effects. \*Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with gait adaptability task.



**Fig 4** Average and SD of  $\lambda_{s-step}$  (A), backward MoS (B), and ML MoS (C) for amputees (white symbols;  $n=9$ ) and healthy controls (black symbols;  $n=9$ ). Abbreviations: NW, normal walking; P, perturbation. +Significant group effects. \*Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with gait adaptability task.

study. Besides, when interpreting the results of the present study, it has to be taken into account that the transtibial amputees who participated in this study were all relatively young and generally good walkers. All amputees, except 1, had amputations performed after trauma. For this group of amputees, the overall walking ability is in general higher than for people with an amputation performed because of vascular disorders.<sup>13</sup> All of these aspects may explain why amputees in this study could adapt their gait pattern relatively well to the applied manipulations, but generalization to less proficient walkers needs to be done with caution. Another limitation of this study is the estimation of the CoM as the average of the markers attached to the LASI, RASI, LPSI, and RPSI of the pelvis to calculate the XCoM and subsequently the MoS in the ML and BW direction. This is not the real representation of the CoM, but errors made were likely similar for both groups across conditions. Therefore, these errors would not affect differences in MoS between both groups and between conditions.

The clinical implication of this study is that not all differences in the gait pattern between people with a lower limb prosthesis and able-bodied people are dedicated to enhancing MoS, but might serve other functional constraints. Nevertheless, people with a lower limb amputation can and do use effective stepping strategies to enhance MoS when confronted with challenges to gait stability and adaptability. Because of the relatively high walking capabilities of the participating amputees, the observations made in this study can be used as a reference in assessments of walking ability of people with a transtibial amputation in more general, as well as in defining training goals for the training of adequate strategies that can be used to enhance stability and adaptability during walking.

## Conclusions

In response to the balance perturbations, both transtibial amputees and able-bodied people increased their ML and BW

MoS by increasing their step frequency and step width and decreasing their step length, while their walking speed did not change with respect to normal walking. To enhance gait adaptability, both groups increased step width and decreased step length to maintain ML and BW MoS, while step frequency was not adapted, to prevent a deterioration of the accuracy of the hitting movement. During unperturbed walking, amputees walked with wider steps to increase the ML MoS, possibly to compensate for a lower LDS. Other deviations of the general gait pattern of transtibial amputees compared with able-bodied people, such as the overall lower step frequency and walking speed, do not seem to contribute to enhanced stability or adaptability and should therefore be attributed to other goals or deficits in prosthetic gait.

## Suppliers

- Motek Medical BV, Keienbergweg 77, 1101 GE Amsterdam, The Netherlands.
- Vicon, 14 Minns Business Park, West Way, Oxford, OX2 0JB, UK.
- SPSS Inc, 233 S Wacker Dr, 11th Fl, Chicago, IL 60606.

## Keywords

Amputees; Walking; Rehabilitation

## Corresponding author

Han Houdijk, PhD, Research Institute MOVE, VU University Amsterdam, Van der Boechorststraat 9, 1081 BT Amsterdam, The Netherlands. E-mail address: [h.houdijk@vu.nl](mailto:h.houdijk@vu.nl).



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